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(54) Title: APPARATUS AND A METHOD FOR FITTING A HEARING AID
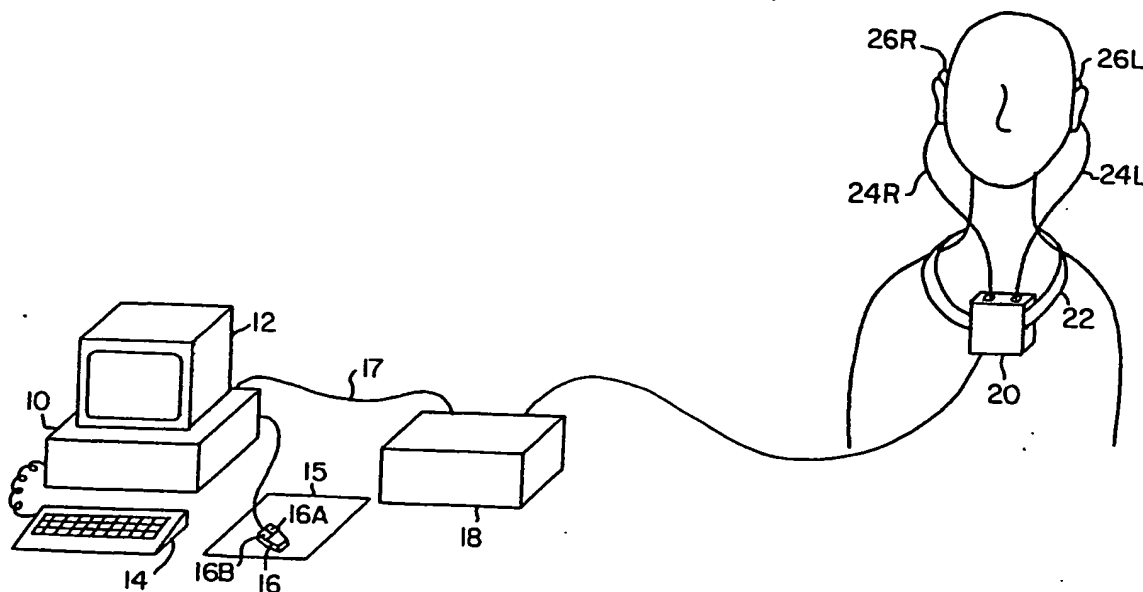
(57) Abstract

A hearing aid fitting system includes a computer having a graphical display which controls apparatus that can program and evaluate an adjustable, multi-band hearing aid. The fitting procedure implemented on this system determines the combined frequency response characteristic of the hearing aid and a client's ear, determines maximum amplification factors which can be implemented in each channel without inducing feedback, and automatically adjusts the hearing aid so that the combination of the client's loss curve, the determined combined frequency response characteristic, and the adjusted hearing aid match a target curve. The target curve and the loss curve may be graphically specified and adjusted using the computer.

-1-

5   APPARATUS AND A METHOD FOR FITTING A HEARING AID

Technical Field

10   The present invention relates generally to
the fitting of hearing aids and in particular to a
computer controlled fitting procedure which allows the
user to interact in the fitting process, using his
particular hearing aid device.
15

Background Art

            Generally, the correction of a hearing
defect proceeds in several steps. First, a hearing
20   test is performed. In this test, an audiometer is
used to determine hearing thresholds for a range of
distinct tones. In the next step, the test results
are analyzed to classify the hearing defect. This
classification may indicate that a particular type of
25   hearing aid would be best for the person having the
defect. The selected hearing aid is then adjusted to

- 2 -

have a frequency response characteristic that corrects
the hearing defect indicated by the audiometric data.
Finally, the hearing test is performed again, this
time with the hearing aid in place, to determine the
5   level of correction that has been achieved.

The process outlined above has several
potential problems. First, there is an undesirable
time delay between the audiometric testing of a client
10   and the final fitting of the adjusted device. The
length of this delay depends on the availability of a
hearing aid that can correct the particular hearing
defect, and the complexity of the required
adjustments. Delays of between of one day and several
15   weeks are not uncommon. These delays are not only a
problem for the client who, obviously, would like to
have his or her hearing aid as soon as possible, but
delays are also a problem for the hearing aid
provider. The provider's problems mainly relate to
20   clients who may refuse to accept a device fitted
several days before, claiming either that the testing
procedure was improper or that the selection and
adjustment of the device were improper.

25   Second, audiometric tests performed using
asychronous receiver-transmitter (UART). instructions
a standard audiometer do not adequately measure
hearing loss and cannot account for the mechanical
response of the ear with a hearing aid in place.

SUBSTITUTE SHEET

- 3 -

Consequently, the prescribed hearing aid may appear to
perform well on the audiometric tests but may
inadequately correct the users perception of other
sounds such as speech. Moreover, a hearing aid that
5   is fitted using only audiometric data cannot
adequately compensate for resonances and nulls that
result from mechanical interaction of the hearing aid
and the ear canal.

10   Finally, in the steps described above, there
is no opportunity for the individual being fitted to
include his or her own preferences in the performance
of the unit until after it has been selected and
adjusted. To accommodate these preferences, the
15   provider may have to make further adjustments to the
device, causing further delays the fitting process.

Description of the Invention
20

The present invention is embodied apparatus
and a method for fitting a person with a programmable
hearing aid. A hearing aid fitted using this method
desirably includes a plurality of separate channels
25   each of which processes signals occupying a distinct
band of frequencies. Data representing a client's
hearing defect and data representing a target
frequency response characteristic to correct the
defect are entered into a computer. Responsive to

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- 4 -

this data, the computer determines adjustment
coefficients for the plurality of separate channels
which most closely match the frequency response
characteristic of the hearing aid to the target
5   characteristic. These adjustment coefficients are
then used to program the hearing aid while it is in
place, coupled to the clients ear.

        In accordance with a further application of
10   this invention, before performing the steps set forth
above, the receiver of a programmable hearing aid and
a separate probe microphone are inserted into a
client's ear canal. An exemplary probe microphone 410
is described below in reference to FIGURE 4. Using
15   external signals applied to the hearing aid and
response signals provided via the probe microphone,
the combined frequency response characteristic of the
hearing aid and the client's ear canal are determined.
Next, using these same signals, a maximum
20   amplification factor is determined for each of the
plurality of channels by separately adjusting the
level of amplification for each channel. Both the
combined frequency response characteristic and the
maximum amplification factors are entered into the
25   computer, with the data representing the client's
hearing defect and the target frequency response
characteristic, to develop the plurality of
coefficient values.

- 5 -

            In accordance with another aspect of this
    invention, an operator may enter data regarding the
    client's hearing defect or data representing client
    preferences in the programming of the hearing aid into
5   the computer directly on a graphic display of this
    data, using a pointing device.

    <u>Brief</u> <u>Description</u> <u>of</u> <u>the</u> <u>Figures</u>

10   Figure 1 is an illustration of major
    elements of the fitting system.

            Figures 2A and 2B are drawings which show
    the apparatus physically fitted to the client during
15 the fitting operation.

            Figure 3 is a block diagram of circuitry
    suitable for use as the interface unit of the system
    shown in Figure 1.
20
            Figure 3A is a flow-chart diagram useful for
    explaining the operation of the interface unit shown
    in Figure 3.

25   Figure 4 is a block diagram of circuitry
    suitable for use as the chest pack of the system shown
    in Figure 1.

Figure 5 is a block diagram of an exemplary
programmable hearing aid which may be fitted using the
system shown in Figure 1.

5   Figure 6 is a schematic diagram, partially
in block diagram form of an attenuator suitable for
use in the hearing aid shown in Figure 5.

Figures 7, 8, 9 and 10 are flow-chart
10   diagrams which illustrate the operation of the system
shown in Figure 1.

Figure 11 is a diagram useful for explaining
the operator interface to the system shown in Figure
15   1.

Detailed Description of the Figures

20   Figure 1 is a pictorial representation of the
major elements of a hearing aid fitting system which
includes an embodiment of the present invention. The
system includes a computer 10 having a video display
12, a keyboard 14 and a pointer-type data entry device
25   16, commonly referred to as a "mouse." The computer
10 may be, for example, an IBM Personal Computer.^{TM} An
operator performing a fitting operation, enters data
and commands for the fitting system into the computer
10 using the keyboard 14 and the mouse 16. The video

display 12 may be used to display audiometric data
relating to the client being fitted.

In the present embodiment of the invention,
a hearing aid is described which includes a 13 band
equalizer. The frequency response characteristic of
the hearing aid is determined by the net gain or
amplification factors of each of the 13 frequency
bands. All amplification factors and attenuation
levels used in this application are in units of
decibels (dB). Consequently, a positive attenuation
level is equivalent to a negative amplification factor
and the net gain of two cascaded amplifiers is
determined by summing their individual gains.

The computer 10 is coupled by a data link 17
to an interface unit 18. The interface unit 18 is
coupled to a chest pack 20, suspended from the neck of
the client by a strap 22, and the chest pack 20 is
coupled, by cables 24R and 24L, to one or two hearing
aid devices 26R and 26L, respectively. Using the
computer 10, interface unit 18 and chest pack 20, an
operator can change the frequency response
characteristic of the hearing aids 26R and 26L; apply
audio signals through the hearing aids 26R and 26L to
obtain audiometric data relating to the client; and,
through probe microphones inserted into the client's
ears, monitor the performance of the hearing aids 26.

        Figures 2A and 2B are two views of a
client's right ear which illustrate apparatus used in
the fitting process. As shown in Figure 2A, the
hearing aid device 26R is physically fitted behind the
5   client's ear with an earmold 212 coupling the receiver
of the hearing aid into the client's ear canal. A
probe microphone, 210R is also inserted into the
client's ear canal and is held in place by the earmold
212. An identical probe microphone 210L (not shown)
10   is inserted into the client's left ear canal. A
signal conducting path, 24A, for applying operating
power and programming signals to the hearing aid 26R,
is coupled via a connector (not shown) which may be
located, for example, in the battery compartment (not
15   shown) of the hearing aid 26R. In addition, a signal
conducting path 24B, conveys signals from the probe
microphone. The conduction paths 24A and 24B are
physically coupled to form a single cable, 24R which
couples the hearing aid 26R to the chest pack 20. The
20   signals conveyed by these conduction paths are passed
through the chest pack 20 as set forth below in
reference to Figure 4, to or from the interface unit
18.

25   Circuitry suitable for use as the interface
unit 18 is shown in Figure 3. Data communication
between the interface unit 18 and the computer 10 is
implemented by a receiver/transmitter unit 310. The
unit 310 may be, for example, a conventional universal

- 9 -

asychronous receiver-transmitter (UART) circuit.
Instructions and data from the computer 10 are passed
through the unit 310 to a microprocessor 312. The
microprocessor 312, using a program stored in a memory

5   314, processes the information received from the
computer 10 to control a programmable signal
generator 320, an analog-to-digital converter (ADC)
322 and a parallel-input serial-output register 316.
The programmable signal generator used in this

10   embodiment of the invention is a 5503 integrated
circuit available from Ensonic Corporation. The bit-
serial output signal of the register 316 is applied to
the chest pack 20, via a synchronous input/output
(I/O) device 318, to control the amplification factors

15   for the 13 signal processing channels of the each of
the hearing aids 26. The I/O device 318 also provides
a synchronizing clock signal, CKPA and a signal, L/R,
to the chest pack 20. The signal L/R selects either
the hearing aid 26L or 26R to receive the programmed

20   amplification factors.

Figure 3A is a flow-chart diagram of the
program that controls the microprocessor 312.
Functionally, this program accepts commands from the

25   computer 10 and based on the type of command received,
requests parametric data from the computer 10. The
microprocessor 312 then uses the requested data to
condition its various peripheral devices to perform
the requested command. When the command has been

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completed, the unit 18 sends a message to that effect,
with any data that may have been requested, to the
computer 10. As an error check, all transmissions
between the computer 10 and interface unit 18 include

5   a checksum. If an invalid checksum is detected, the
receiving unit, 10 or 18, requests the originating
unit to retransmit the data.

The first step, 350, in Figure 3A waits for

10   a command from the computer 10. When a command is
received, the microprocessor 312, at step 352
determines the command type. After processing any of
the commands listed below, the microprocessor 312
branches back to the step 350 to await the next

15   command.

If, at step 352, the command is a request to
change the amplification factor or gain of a signal
processing channel of one of the hearing aid devices

20   26R or 26L, the microprocessor 312, at step, 354,
requests parametric information from the computer 10 to
selects a hearing aid device 26L (left) or 26R
(right), a channel (1 to 13) and a new attenuation
level (0 to 63) for a programmable attenuator in the

25   selected channel. The next step in the program, 355,
sends a signal MUTE to the chest pack 20. This signal
mutes the selected hearing aid while the attenuation
level is changed.

Using the attenuator and gain selection
data, the microprocessor 312, at step 356, loads 13
six-bit values representing the desired attenuation
levels for the selected hearing aid 26R or 26L into
5   the shift register 316. At step 358, the
microprocessor 312 conditions the synchronous I/O unit
318 to apply the 13 six-bit values to the chest pack
20 as a bit-serial signal, PA. The I/O unit 318 also
provides a clock signal CKPA to synchronize the chest
10   pack 20 to the signal PA and a signal L/R to indicate
whether the data is for the hearing aid 26L or 26R.

As an error check, the chest pack 20
automatically retransmits the data it receives to the
15   interface unit 18. This data is received at step 360,
via the synchronous I/O unit 318, which, in turn,
provides the data to the microprocessor 312. The
microprocessor 312 checks the received data against
internally stored attenuation levels for the selected
20   hearing aid. If an error is detected, the program
branches back to step 358 and attempts to retransmit
the data. If no error is detected, step 364 is
executed which changes the state of the signal MUTE to
reverse the muting operation performed in step 355
25   above. The effect if the signal MUTE on the hearing
aids 26 is described below in reference to Figure 5.

When the final attenuation levels have been
applied to the hearing aids 26, they may be stored

- 12 -

permanently into an electrically erasable read only
memory (EEPROM) internal to the hearing aids 26. This
command is conveyed to the chest pack 20 via a signal
BURN provided by the microprocessor 18.

5

Another type of command which may be issued
by the computer 10 is a request that a signal having a
particular frequency and amplitude be sent to a
selected one of the hearing aid devices 26. If a
10   command of this type is detected at step 352, the
microprocessor 312, at step 370, requests parameters
from the computer 10 that select the left or right
hearing aid and that define the frequency and
amplitude of the signal to be sent. At step 372, the
15   signal generator is conditioned to provide the
requested signal. At step 374, a demultiplexer 321 is
then conditioned by the microprocessor 312 to select a
path to the right or left hearing aid 26R or 26L
through the chest pack 20. As set forth below in
20   reference to Figures 4 and 5, this audio signal is
passed through the chest pack and applied internally
to the selected hearing aid 26, bypassing the
microphone input to the hearing aid. The parameters
provided by the computer 10, through the
25   microprocessor 312, may condition the signal generator
320 to provide different types of sound such as
continuous tones, discontinuous tones or narrow band
noise, and to provide sound signals having different
durations.

30

The last type of command for this exemplary
embodiment of the invention is a request to receive
data from one of the probe microphones 210R or 210L,
in the client's ear canals. The first step for this
5   command, 380, obtains a parameter from the computer 10
which selects the probe microphone for the left or
right hearing aid to provide the data. At step 382,
the microprocessor 312 conditions a multiplexer 323 to
select the signal provided by the selected hearing
10   aid. The selected signal is applied to an RMS
converter 324 which develops a direct current (DC)
signal representing the sound pressure inside the
client's ear canal. This signal is applied to the ADC
322 which, at step 384, digitizes the signal and
15   applies it to the microprocessor 312. The
microprocessor 312, at step 386, formats the data for
transmission to the computer 10 at step 388.

Figure 4 is a block diagram illustrating
20   circuitry suitable for use in the chest pack 20. In
the chest pack 20, a shift register 412 is coupled to
receive the signal PA provided by the interface unit
18. As set forth above, the signal PA is sent back to
the interface unit 18 for error correction and
25   detection immediately after it has been received.
the signal conduction lines from

After the error checking procedure, the

stored values in the shift register 412 are applied to
level shifting circuitry 414 which converts the
signals from logic levels used by the interface unit
18 to lower logic levels used by the hearing aid

5   circuitry. After this conversion, the signals are
applied to the selected hearing aid 26R or 26L via a
demultiplexer 416, as directed by the signal L/R
provided by the interface unit 18.

10   The audio signals, SIGL and SIGR provided by
the interface unit 18 are applied to the respective
left and right hearing aid devices 26R and 26L. The
signals 24A generated by the probe microphones 210 of
the respective hearing aids 26R and 26L are amplified

15   by the respective amplifiers 422 and 424, in the chest
pack 20, to generate the signals MIC_R and MIC_L which
are applied to the interface unit 18. On the hearing
aid end of the chest pack 20, all of the signals for
the left hearing aid 26L are grouped together to form

20   the cable 24L and all of the signals for the right
hearing aid are grouped together to form the cable
24R. As shown in Figure 1, these cables are applied
to the respective hearing aid devices 26R and 26L.
The signal MUTE and the signal CK_{PA} are passed through

25   respective level shifting circuits 418 and 420 which
adjust the logic levels produced by the microprocessor
°[312 and I/O unit 318 to the +- 1.25 volt logic levels
used in the hearing aids 26. The signal BURN is also
passed through a level shifter 419, however, this

- 15 -

level shifter changes the signal BURN to a level
sufficient to "burn" data into the EEPROMs of the
hearing aids 26.

5   The chest pack 20 receives its operational
power from the interface unit 18 and also provides
operational power to the hearing aid devices 26R and
26L. This power supply circuitry is not shown to
simplify the description of the invention.

10

          Figure 5 is a block diagram circuitry
suitable for use as either of the hearing aids 26R or
26L. This circuitry includes thirteen distinct signal
processing channels which process signals in thirteen
15   respective frequency bands. Each channel includes a
band-pass filter and an programmable attenuator. The
programmable attenuators may be adjusted, using
digital interface circuitry, so that the hearing aid
may be configured to compensate for a wide variety of
20   hearing deficiencies. In addition, the five lowest
frequency channels may be completely disabled in the
presence of relatively high levels of ambient noise,
both to save power by not processing this noise and to
increase the intelligibility of words spoken over the
25   ambient noise.

          The hearing aid shown in Figure 1 includes a
VLSI integrated circuit (IC) 500 which uses the power-
efficient complementary metal oxide semiconductor

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(CMOS) technology. Further power efficiency is gained
by using switched-capacitor designs for the band-pass
filters and programmable attenuators.

5   In the description set forth below, the
circuitry used in this hearing aid is generally
described in terms of its function. This general
description is followed by a more detailed description
of the programmable attenuators used in the various
10   signal processing channels.

            In Figure 5, sound waves are converted to
electrical signals by a microphone 510, which applies
the electrical signals to a compression amplifier 512.
15   The amplifier 512 amplifies relatively quiet sounds
and attenuates relatively loud sounds so that the
signals provided by the amplifier remain substantially
within a predetermined dynamic range. This dynamic
range is compressed relative to the original dynamic
20   range. The circuitry 512 amplifies signals in a
frequency range from 100 Hz·to 10 KHz. The output
signals of the compressor amplifier 512 are applied to
a further amplifier 514 which drives the 13 signal
processing channels. The amplifiers 512 and 514 used
25   in this embodiment of the invention provide a net
amplification factor of 56 dB. These amplifiers are
available as a single integrated circuit, the LD-512
manufactured by Gennum Corp.

As set forth above, each of the 13 signal
processing channels includes a band pass filter and a
programmable attenuator. The band-pass filters used
in the respective signal processing channels are

5   numbered 516 through 528 and their corresponding
attenuators are numbered 540 through 552. The band-
pass filters used in this embodiment of the invention
are of conventional design. As shown in Figure 5, the
band pass filter 516 has a one octave passband, the

10   filters 517 and 518 have one-half octave passbands and
the filters 520 through 528 have one-third octave
passbands. The center frequencies for the filters 516
through 528 are 178 Hz, 338 Hz, 588 Hz, 776 Hz, 1.04
KHz, 1.44 KHz, 1.8 KHz, 2.3 KHz, 2.9 KHz, 3.6 KHz, 4.6

15   KHz, 5.8 KHz, and 7.2 KHz. The transition regions of
the frequency response characteristics of these
exemplary filters exhibit a roll-off of approximately
12 dB/octave. These frequency response
characteristics were chosen so that the 13 channels

20   convey approximately equal amounts of speech
information. Circuitry suitable for use as one of
these switched-capacitor band-pass filters is
disclosed in U.S. Patent 4,622,440 entitled
"Differential Hearing Aid With Programmable Frequency

25   Response" which is hereby incorporated by reference.

Each of the thirteen band-pass filters
passes signals in a respectively different band of
frequencies. However, because each of the filters

passes signals outside of its passband at a reduced
amplitude, there may be significant amounts of overlap
in the signals passed by successive filters.

5   Each of the attenuators 540 through 552 is
separately programmable via the chest pack 20 which
provides programming signals, as set forth above, to a
digital interface 92 and an electronically erasable
programmable read-only memory (EEPROM) 94. Each of
10   the attenuators 540 through 552 may be programmed to
provide up to 40 dB of attenuation to the signals
applied to its input terminal. An exemplary
attenuator is described below in reference to Figure
6.
15

          The output signals provided by the
attenuators 540 through 552 are applied to summing and
sample-and- hold circuitry 560. The circuitry 560 adds
together all of the signals provided by the
20   attenuators and samples the summed signal to remove
any sampling artifacts introduced by the switched
capacitor band- pass filters 516 through 528 and
attenuators 540 through 552.

25   The output signal of the circuitry 560 is
applied to an amplifier 570. The amplifier 570 is
responsive to a volume control 571 to allow the user
to adjust the level of sound produced by the hearing
aid. During the fitting procedure, the volume control

- 19 -

571 is set to a predetermined position. The amplifier
570 provides an amplification factor between 0 dB and
40 dB. The signals produced by the amplifier 570 are
applied to a final amplifier 572 which provides an
5   amplification factor of 10 dB. In this embodiment of
the invention, the final amplifier 572 is not a part
of the IC 500. The output signals of the final
amplifier 572 are applied to a final attenuator 574
which is a part of the IC 500.
10

The final attenuator 574 includes three
resistors, RA, RB and RC and four switches, 574A,
574B, 574C and 574D. Each of these switches is
controlled by a separate bit provided, in normal
15   operation, by the EEPROM 594 or, during the fitting
procedure by the chest pack 20 via the digital
interface 592. When the switch 74D is closed, the
output of the final amplifier 572 is coupled directly
to the receiver 576, so there is no attenuation. If
20   the switch 574D is open and any of the other switches
574A, 574B and 574C are closed, a resistance is
inserted in the path between the final amplifier 572
and a receiver 576. The amount of attenuation
provided by any combination of resistors depends on
25   the relative impedances of the combined resistors and
the receiver. The final attenuator 574 provides
between 0 and 40 dB of attenuation. The amount of
attenuation provided is adjustable via the computer
10, interface unit 18, chest pack 20, digital

- 20 -

interface 592 and EEPROM 594. This attenuator may be
programmed in the same manner as the attenuators 540
through 552 except that the code for the different
attenuation levels would be between 0 and 8 instead of
5   between 0 and 63. The output signal of the final
attenuator 574 drives the receiver 576 which produces
sound waves in the ear of the user.

An output signal 513 of the compressor
10   amplifier 512, which indicates the sound pressure
level at the microphone 510, is applied to an ambient
sound compensation circuit 515. The circuit 515
detects ambient sound levels having amplitudes in a
predetermined range. Responsive to these detected
15   levels, the circuit 515 selectively disables up to
five of the lower frequency channels of the hearing
aid circuitry using switches 529 through 533. This
step successively disables the low frequency channels
in the presence of high levels of ambient noise, both
20   to save power and to increase the intelligibility of
speech. Power is saved since the input signals to the
band-pass filters and attenuators for the low
frequency channels are disabled when high levels of
signal are present. The intelligibility of speech is
25   increased because the low frequency bands, which
generally contain a large portion of the ambient
noise, are deemphasized relative to higher frequency
bands which contain a relatively large portion of the
speech information.
30

- 21 -

Figure 6 is a schematic diagram, partially
in block diagram form, of circuitry suitable for use
as any of the programmable attenuators 540 through
552. This attenuator includes an operational
5   amplifier 610 having a fixed capacitor 611 and a
switched capacitor network 612 in a feedback loop from
the output terminal, O, of the operational amplifier
610 to its inverting input terminal. In addition the
attenuator includes six switched capacitor networks,
10   614, 616, 618, 620, 622 and 624 which are coupled
between an input terminal, I, of the attenuator and
the inverting input terminal of the operational
amplifier 610. Each of the switched capacitor
networks 614, 616, 618, 620, 622 and 624 includes a
15   switch in its input path. These switches, 628, 630,
632, 634, 636 and 638 are controlled by signals G1,
G2, G3, G4, G5 and G6, respectively, to selectively
apply input signals provided via the input terminal I,
through their respective switched capacitor networks,
20   to the inverting input terminal of the operational
amplifier 610.

The six signals G1 through G6 are the six-
bit attenuation control signal provided by the digital
25   interface 592. These signals are inverted by the
respective inverters 640, 642, 644, 646, 648 and 650
to provide both inverted and noninverted versions of
the signals to control the respective switches 628,

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630, 632, 634, 636 and 638. Each of these switches and each switch in each of the switched capacitor networks may be, for example, a conventional CMOS transmission gate.

5

The switched capacitor networks 612, 614, 616, 618, 620, 622 and 624 are each responsive to two antiphasal clock signals, C1 and C2. These clock signals may have a frequency of, for example, 100 KHz

10   and a duty cycle of slightly less than 50%. Except for the value of their capacitances, all of the switched capacitor networks are identical. The description of the structure and operation of the network 612, set forth below applies to the networks

15   614, 616, 618, 620, 622 and 624, which, for the sake of brevity, are not described in detail.

The switched capacitor network 612, includes four switches, S1, S2, S3 and S4 which alternately

20   couple a capacitor 613 between the output terminal O and a source of reference potential (e.g. ground) on the one hand and the inverting input terminal of the operational amplifier 610 and ground on the other hand. The clock signal C1, which controls the

25   switches S1 and S4, is substantially antiphasal to the signal C2 which controls the switches S2 and S3, so, when the switches S1 and S4 are open, the switches S2 and S3 are closed and vice-versa.

30

        In a first time interval, when the switches
S2 and S3 are closed, terminal A of the capacitor 613
is coupled to ground and terminal B is coupled to the
output terminal O of the operational amplifier 610.
5   During this time interval, the current provided by the
amplifier 610 charges the capacitor 613 to a potential
which depends on the amount of current provided by the
amplifier 610, the capacitance of the capacitor 613
and the length of the first time interval. At the end
10   of the first time interval, the switches S2 and S3 are
opened and the switches S1 and S4 are closed coupling
terminal A of the capacitor 613 to the inverting input
of the amplifier 610 and terminal B to ground for a
second time interval. In this configuration, the
15   capacitor 613 is coupled to apply the inverse of the
potential developed during the previous time interval
to the inverting input of the amplifier 610. This
potential is summed with potentials provided by the
switched capacitor networks 614, 616, 618, 620, 622
20   and 624 to provide an input potential to the amplifier
610.

        The fixed capacitor 611 configured in
parallel with the switched capacitor network 612 acts
25   to stabilize the amplifier 610 and to remove any high
frequency artifacts of the switched capacitor
processing. This capacitor conditions the circuitry
shown in Figure 6 to operate as a low-pass filter
having a cut-off frequency that is above the highest

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- 24 -

audio frequency processed by the hearing aid but below
the sampling frequency of the switched capacitor
networks, that is to say, below the frequency of the
clock signals C1 and C2. Thus, artifacts of the
5   switched capacitor processing are rejected by the
attenuator circuitry shown in Figure 6.

It is well known that switched capacitor
networks such as 612, 614, 616, 618, 620, 622 and 624
10   may be modeled as resistors in analyzing the
performance of the circuit. The value of the
capacitor in a network is inversely proportional to
the equivalent resistance value for the network. If
each of the switched capacitor networks in the
15   circuitry shown in Figure 6 were replaced by its
equivalent resistance, the circuitry would be
configured as a conventional class A amplifier, the
gain of the amplifier being determined by the ratio of
the feedback resistance to the input resistance.
20

In the circuitry shown in Figure 6, the
values of the capacitors in the switched capacitor
networks are chosen such that the gain of the
amplifier is always fractional and so, the circuitry
25   shown in Figure 6 always attenuates signals applied to
its input terminal. The capacitors are selected such
that the network 614 provides 20 dB of attenuation,
and the networks 616, 618, 620, 622 and 624 divide up
an additional 20 dB of attenuation into 0.5 dB steps.

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Thus the attenuator shown in Figure 6 may be programmed to provide signal attenuation in a stepwise continuous range between approximately 0.5 dB and 40 dB.

5

The description set forth above relates to the apparatus used to fit the hearing aids 26 to a client. The actual procedure used in the fitting operation is described below. Since the apparatus

10 described above is all controlled directly or indirectly by the computer 10, the fitting procedure is described in terms of a flow-chart diagram for a program implemented on the computer 10. In the present embodiment of the invention, this program is

15 written in the C computer programming language. In addition, the program includes an interface to the Microsoft (tm) Windows program to provide the operator with a means for graphically entering commands and data.

20

Figure 7 is a flow-chart diagram which illustrates the overall procedure for fitting one hearing aid device. If a client is to be fitted with two hearing aids, each step shown in Figure 7 is

25 performed twice, once for the right ear and once for the left.

The first step in Figure 7, step 710, calibrates the hearing aid device. This step, which

- 26 -

is described in greater detail with reference to
Figure 8, determines the combined frequency response
characteristic of the hearing aid and the client's ear
canal. The next step, 712, finds the maximum

5   amplification factor which may be realized in each of
the 13 signal processing channels. The useful
amplification factor for a particular channel or
channels may be limited by positive feedback between
the receiver 576 and microphone 710. The procedure

10   for determining the maximum amplification factor is
described below in reference to Figure 9.

In step 714, the hearing loss curve for the
ear being fitted is specified. This step may be

15   performed, for example, by having the operator use the
computer 10 to plot a previously determined loss curve
on a fitting work sheet as set forth below in
reference to Figure 11. Alternatively, client's loss
curve may be determined by performing standard

20   audiometric tests, using the signal generator 320 of
the interface unit 18 and the hearing aids 26 as an
audiometer.

After the client's loss curve has been

25   established, the operator, at step 716, specifies a
target correction curve. This curve represents the
desired corrected response for the client. The target
curve may be entered manually on the fitting work
sheet or determined automatically using one of several

SUBSTITUTE SHEET

standard formulas such as the Berger, POGO or N. A. L.
formulas. The Berger formula is set forth in a paper
by K. Berger et al. entitled "A Method of Hearing Aid
Prescription", Hearing Instruments, July, 1978. The

5   POGO formula is described in a paper by G. McCandless
et al. entitled "Prescription of Gain/Output (POGO)
for Hearing Aids", Hearing Instruments Vol. 34, No. 1,
1983. The N.A.L. formula is described in a
publication by D. Byrne et al. entitled "The National

10   Acoustic Laboratories' (NAL) New Procedure for
Selecting the Gain and Frequency Response of a Hearing
Aid". This publication is available from National
Acoustic Laboratories, Chatswood, N.S.W., Australia.

15   In step 718, the computer 10 adjusts the
amplification factors for each of the 13 signal
processing channels to match the combined response,
represented by the loss curve augmented by the hearing
aid (i.e. the aided response curve), to the target

20   curve. This step, which is described in detail below,
with reference to Figure 10, calculates the best set
of attenuation levels for the 13 signal processing
channels of the hearing aid device and then applies
these attenuation levels to the device. This

25   procedure may not be able to exactly match the target
curve due to limitations imposed by the combined
frequency response characteristic of the hearing aid
and the client's ear and by the maximum amplification
factors that can be set in each channel as determined

- 28 -

at step 712.

        After the hearing aid has been adjusted to
implement the calculated amplification factors, a step

5   720 is executed to determine if further adjustment is
desirable. An adjustment of this type may, for
example, attempt to correct for differences between
the actual performance of the hearing aid and its
calculated performance, as determined by audiometric

10   tests performed with the adjusted hearing aid in
place. A further adjustment may be made, simply to
accommodate a client's individual preferences, such as
more high frequency or low frequency in the audio
output signal. In response to these tests or

15   preferences, the operator, at step 722, adjusts the
target curve on the work sheet. The program then
branches to step 718 to change the amplification
factors of the 13 channels to match the new target
curve. When, at step 720, it is determined that no

20   further adjustment is desirable, the step 724
terminates the fitting procedure. This step may
involve, for example, recording the client's name and
address as well as the various response curves and
amplification factors determined during the fitting

25   procedure for future reference.

        Figure 8 is a flow-chart diagram of the
procedure used to calibrate one of the hearing aids
26R and 26L. The first step, 810, sets all of the

attenuators to a known attenuation level. This level
may represent an attenuation level of 0 dB, 40 dB or
any other value in-between. Next, at step 812, the
variable CHANNEL is set to zero. At step 814, the
5   variable CHANNEL is incremented to select the first
channel. In the next step, 816, the computer 10
conditions the interface unit 18 to apply a sound
signal to the selected hearing aid. This sound signal
has an amplitude level that will not cause any
10   discomfort to the client and has a frequency centered
in the pass-band of the selected channel. It
represents an acoustic signal having a known sound
pressure level at the microphone input to the hearing
aid.
15

While this signal is applied to the selected
hearing aid, the computer 10, at step 818, requests
the interface unit 18 to provide a measurement of the
sound pressure in the client's ear canal, that is to
20   say, a measure of the acoustic signal produced by the
hearing aid in the client's ear. At step 820, the
sound pressure level provided by the interface unit 18
is divided by the known input sound pressure level and
the result is converted to an amplification factor in
25   decibel units. The value produced by this operation
represents the amplification factor of the combination
of the hearing aid and the client's ear canal at the
selected frequency. This value is stored by the
computer 10, step 822, for use in setting attenuation

SUBSTITUTE SHEET

levels for the 13 channels of the selected hearing
aid. At step 824, the computer 10 branches back to
step 814 if the selected channel is less than 13 and
branches to step 826, ending the calibration process
5   otherwise.

    Figure 9 is a flow-chart diagram of the
program that determines the maximum amplification
factors that should be implemented in each of the 13
10   channels. This program is step 712 of Figure 7. If
the amplification factor for a particular channel is
too great, the acoustic signal provided by the
receiver in the client's ear may be picked up at the
microphone input of the hearing aid, causing the
15   hearing aid to oscillate, producing a loud ringing
noise. This noise is commonly known as feedback.

    The process set forth in Figure 9 analyzes
each channel separately and determines the largest
20   amplification factor for the channel which will not
produce feedback. This amplification factor is
recorded by the computer 10 and used as an upper limit
when the hearing aid is programmed to match the target
curve. The interface unit 18 does not mute the
25   selected hearing aid during this procedure, so that,
referring to Figure 5, sound signals produced by the
receiver 576 may be picked up by the microphone 510.

    Step 910 of the program shown in Figure 9

sets the attenuators in all 13 channels to their
maximum attenuation levels, that is to say, their
minimum amplification factors. Step 912 initializes
the variable CHANNEL and step 914 increments it,

5   selecting the first channel. In step 916, the
computer 10 conditions the interface unit 18 to reduce
the attenuation level (i.e. increase the amplification
factor) in the selected channel. Next, at step 918,
the interface unit 18, under control of the computer

10   10, applies a narrow-band noise signal to the selected
hearing aid. This signal has a very short duration, a
frequency centered in the pass-band of the selected
channel and an amplitude substantially equal to the
maximum amplitude produced by the compressor amplifier

15   512 of Figure 5.

     At step 920, the computer 10 conditions the
interface unit 18 to monitor the signal received at
the probe microphone after the narrow band noise

20   signal has stopped. If a significant signal is
measured at step 922, the hearing aid is assumed to be
in an oscillatory state. In this instance, at step
924, the attenuation level is incremented by one for
the selected channel to stop the feedback oscillation.

25   The amplification factor corresponding to this
incremented attenuation level is recorded at step 928
as the maximum amplification factor for the
frequencies corresponding to the selected channel.

- 32 -

        If at step 922, no significant signal is
measured, then there is no feedback at the selected
attenuation level. The step 926 determines if the
current attenuation level for the selected channel is
5   the lowest possible attenuation level. If so, the
.step 926 transfers control to the step .928 to mark the
amplification factor corresponding to this attenuation
level as the maximum amplification factor for the
frequencies corresponding to the selected channel and
10   to reset the attenuation level for the selected
channel to its maximum value. Otherwise, the step 926
transfers control to step 916 to further reduce the
attenuation level until either the minimum attenuation
level is attained or feedback oscillation is detected.
15

        Step 930, which is executed after step 928,
checks the selected channel number against the maximum
channel number, 13. If the selected channel is less
than 13, control is transferred to step 914 to select
20   the next channel. Otherwise, step 932 is executed
ending the process.

        When the maximum amplification factors for
each of the 13 channels have been established, the
25   operator specifies the client's loss curves and target
curves and then attempts to correct the client's
hearing loss by adjusting the frequency response
characteristic of the hearing aid. This procedure is
set forth above in reference to Figure 7.
30

- 33 -

Figure 10 is a flow-chart diagram of the
program that adjusts the frequency response
characteristic of a hearing aid 26R or 26L.   This
program corresponds to the step 718 of Figure 7.   The
5   steps 1010, 1012 and 1014 of Figure 10, generate a
table of amplification factors for the hearing aids
26.   The table has 13 columns representing the
respective 13 signal processing channels, and the 64
rows representing the 64 respective attenuation levels
10   for each channel.   Each entry in the table represents
an amplification factor provided by a channel of the
hearing aid represented by the column number of the
entry using an attenuation level represented by the
row number.

15

In step 1010, a table of nominal
amplification factors is derived.   Each nominal
amplification factor is generated from a common base
factor.   Referring to Figure 5, this base factor may
20   be produced by summing fixed amplification factors for
of the amplifiers 512, 514, 570 and 572 and then
subtracting a predetermined fixed attenuation level
for the variable attenuator 574.   To generate the 64
nominal amplification factors for each of the 13
25   channels, each of the 64 variable attenuation levels
for the channel are subtracted from the base
amplification factor.   This table may be generated,
for example, by repeatedly invoking a FORTRAN

SUBSTITUTE SHEET

- 34 -

subroutine, included in Appendix A, once for each attenuation level in each channel.

5   In step 1012, the nominal amplification factors are combined with the calibration data produced by the algorithm set forth in Figure 8. In this step, the calibration response value, in decibels, for each channel is added to each of the 64 entries each row. In step 1014, entries in each

10   column of the table representing amplification factors greater than the respective 13 maximum amplification factors determined by the algorithm set forth in Figure 9 are set to zero.

15   In steps 1016, 1018 and 1020 the computer 10 develops a first approximation of a frequency response characteristic which will correct the client's hearing loss to the response represented by the target curve. To generate this first approximation, the computer 10,

20   at step 1016, determines the difference between the target curve and the loss curve at each of 13 frequencies, the center frequencies of the 13 signal processing channels. If, at step 1018, the difference value is determined to be less than the minimum or

25   greater than the maximum amplification factor in the column, the respective minimum and maximum values are substituted for the difference value. At step 1020, the computer 10 searches each column of the table for the respective difference value determined in step

1016. The row number for each found entry represents
the attenuation level applied to the respective
channel.

5   The sum of the approximated frequency
response characteristic defined by the attenuation
levels provided by step 1020 and the loss curve is
not a very good approximation of the target curve
because the calculation of the amplification factor
10   for any one channel of the hearing aid does not take
into account contributions from the 12 other channels.
To determine how good the approximation is, step 1022
computes the actual frequency response characteristic
of the hearing aid by combining the contribution of
15   all 13 channels for each of the 13 center frequencies.
Two computer programs, written in the language
FORTRAN, which implement this algorithm are included
below in Appendices A and B.

20   The 13 values defining the frequency
response characteristic of the hearing aid are added,
at step 1023, to 13 corresponding values taken from
the loss curve to produce an approximation of the
corrected response.
25

          The next step, 1024, subtracts 13 values of
the target curve, representing the target response at
the 13 center frequencies, respectively, from the 13
corrected response values produced in step 1023. This

SUBSTITUTE SHEET

- 36 -

operation produces a set of 13 error values which
represent the difference between the approximate
response curve and the target curve. In the step
1026, these error values are subtracted from error
5   values produced in a prior iteration through the
algorithm. For the first iteration, the prior error
values are equal to zero.

If the difference between the present and
10   prior error values is greater than a threshold, a step
1027 transfers control to step 1028. In step 1028,
the computer 10 subtracts the present error values
from the respective difference values provided by step
1016, above, to produce a new set of 13 difference
15   values. Next, at step 1018, these difference values
are limited and at step 1020, the amplification factor
array is searched for the new difference values. This
process continues through steps 1022, 1024 and 1026
and back to step 1018 until the step 1026 determines
20   that the difference between the current and prior
error values is less than the threshold. In this
instance, step 1026 transfers control to step 1030 to
end the algorithm.

25   Although, in the present embodiment of the
invention, step 1026 only tests the current and prior
error values against a threshold, it is contemplated
that an additional test may be performed, one which
branches to step 1030 if the number of iterations

SUBSTITUTE SHEET

through the filter adjusting algorithm is greater than
a predetermined maximum value. This test would
terminate the filter adjusting algorithm when the
error values produced by step 1026 oscillates between
5   two values, one representing an overshoot of the
target curve and one representing an undershoot.

The material set forth above describes the
apparatus used in the fitting procedure and the
10   control of that apparatus through the computer 10.
The following describes how the computer 10 is
controlled by the operator. As described above, in
reference to Figure 1, the operator uses a pointer-
type input device, such as the mouse 16, to enter
15   commands and data via the display device 12, using a
graphic image such as that shown in Figure 11. In
the present embodiment of the invention, this control
mechanism is accomplished using a set of programs
known as the Microsoft (R) Windows package, aviailable
20   from Microsoft Inc. Using this package, a system
designer may produce a graphic image, designate
locations on the image as command entry points and
regions of the image for data entry.

25   For the image shown in Figure 11, a control
box 1112 is provided for entering commands to the
system and a graph of frequency vs. hearing loss, 1110
is provided for entering and displaying data. The
mouse 16 which is used to control the system is

SUBSTITUTE SHEET

represented on the by an arrow 1122 on the displayed
image. When the operator moves the mouse 16 on the
surface of a mouse pad 15, the arrow 1122 moves around
the image in a corresponding manner.

5

     The various boxes and circles in the control
box 1112 are command points that allow the operator to
establish and adjust a loss curve 1114 and a target
curve 1118 for the client's left or right ear and to

10   command the computer 10 to adjust the 13 channel
attenuation factors of the selected hearing aid to
generate an aided response curve 1120. In developing
the aided response curve, the computer 10 makes use of
a maximum gain curve, 1116 which is automatically

15   generated during system initialization as set forth
above in reference to Figure 9.

     In general, the operator uses a command
point by moving the mouse 16 on the mouse pad 15 to

20   position the tip of the arrow 1122 over the desired
command point. The command point is activated by
pressing one of the mouse buttons 16A or 16B. When a
command point allowing the adjustment or specification
of one of the curves on the graph 1110 is activated,

25   the mouse 16 may then be used to position the arrow on
the graph 1110 to perform the change the curve.

     The following example illustrates the use of
the of the mouse 16 and the graphic display shown in

Figure 11 in the context of a typical fitting
procedure. The first step in the procedure is to
physically fit the client with left and/or right
hearing aids 26 including earmolds 212 and to insert
5   the probe microphones 210 into the client's ear
canals. Next, the computer 10, interface unit 18 and
chest pack 20 are initialized. During this step, the
hearing aids 26 are calibrated to the client's ears
and data for the maximum amplification factor curves
10   1116 are developed.

        After initialization, a screen may be
displayed that is similar to the one shown in Figure
11 except that the graph portion 1110 will have no
15   curves. The operator then positions the arrow 1122
over the control point box 1130, marked "Right," and
presses the button. In response to this selection, an
"x" is displayed in the box 1130 and the maximum
amplification factor 1116 for the right hearing aid
20   26R is displayed on the graph 1110.

        Next, the operator specifies the loss curve.
In this example, an air conduction (AC) loss curve was
determined in a previous hearing test so the operator
25   activates the box 1132 in the row marked "AC" and in
the column marked "UnAided" and then activates the
circle marked "S" in the same row. With these command
points activated, the operator moves the arrow 1122 to
the graph 1110, positioning it successively over the

– 40 –

points 1114A through 1114E and pressing one of the
mouse buttons 16A or 16B to establish points for the
curve. When all of the points have been specified,
the computer 10 interpolates interstitial points to
5   generate the displayed curve 1114. By activating the
corresponding command points in the row marked AC(M)
audiometric data developed for one ear using air
conduction signals with masking noise applied to the
other ear may be specified in addition to or in place
10   of the air conduction data.

Alternatively, the loss curve may be
determined by an air conduction or air conduction with
masking audiometric testing procedure (not shown)
15   using the hearing aids 26 to provide sound signals to
the client's ears. At the conclusion of this
audiometric testing, the loss curve 1114 would be
stored by the computer 10. When the control point
1132 is specified, this curve would automatically
20   appear on the graph 1110.

The next step in the fitting procedure is to
generate the target correction curve 1118. This step
is performed by activating the box 1136 marked
25   "Target", then repeatedly pressing the mouse button
while the arrow 1122 is positioned over the box 1138
until the desired calculation method appears in the
box 1140. Finally, the arrow is positioned over the
circle 1142 and the mouse button is pressed to

- 41 -

activate the circle 1142, conditioning the computer 10
to calculate data for the target curve 1118.

The last step in the procedure is to
5   calculate the aided loss curve 1120. This curve
represents the best match to the target curve 1118
which can be achieved adjusting the hearing aid 26R.
This process is initiated by activating the box 1143.
The result of this process is the curve 1120.
10

After performing the initial fit operation,
the hearing aid 26R may be further adjusted by using
the mouse to change the shape of the target curve 1118
and then activating the box 1143 to attempt to fit the
15   client's corrected hearing curve 1120 to the adjusted
target curve. The target curve is adjusted in several
steps. First, the command point 1136 is activated,
then the arrow is moved so that its tip covers a point
on the curve 1118. This point may be moved vertically
20   on the graph 1110 by pressing a mouse button - at
which point the arrow becomes a cross-hair - and,
while the button is pressed, moving the mouse to move
the cross-hair to the desired position. When the
mouse button is released, the curve 1118 is redrawn to
25   pass through each point of the adjusted target curve
including the point that has just been moved.

The new curve, which passes through all of
the specified points, is calculated as follows.

- 42 -

Coordinates representing the first three specified
points for the curve are read from the display. These
coordinates are applied to a polynomial fitting
algorithm which attempts to find a polynomial of
5   degree three or less which passes through the
specified points. However, only the portion of this
polynomial that passes between the first two points is
used. This process is repeated using the previous
second and third points as new first and second
10   points, respectively, and using the next specified
point on the curve 1118 as the new third point. When
the final three specified points are evaluated, the
curve generated by the polynomial fit algorithm is
used for the segment of the response curve 1118
15   spanning all three points. This same algorithm may be
used to calculate any of the curves which may be
specified using the mouse 16 as set forth above.

        At any point when a corrected loss curve
20   1120 is available, the hearing aid 26R may be
temporarily programmed to realize that response. The
response may then be evaluated by the client and may
be adjusted, as set forth above, to accommodate his
preferences. When the final adjustments have been
25   made, they may be permanently programmed into the
nonvolatile EEPROM 594 of the hearing aid. However,
since the EEPROM is electronically erasable, the same
hearing aid may be re-adjusted at a later date to
correct errors in the original fitting procedure or to

- 43 -

track a deteriorating condition in the client.
Although the present invention has been described in
terms of an exemplary embodiment, it is contemplated
that it may be practiced as outlined above with
5   modifications within the spirit and scope of the
appended claims.

-44-

APPENDIX A   Page 1

```
C     THIS PROGRAM CALCULATES THE MAGNITUDE AND PHASE ANGLE
C     OF A SINGLE BAND-PASS FILTER
C
C     THE GAIN CAN BE SET BY SCALING THE MAGNITUDE OF THE
C     OUTPUT
C
C
      REAL*4 FREQ,GAIN,RE,IM,PHASE,FREQ1,FREQ2,DB
      REAL*8 WT,W,T,FQ,POWER,CK0,CK1,CK2,CK3,C3,C4,K0,K1,
     *K2,K3
      INTEGER K,L,LQ
      COMPLEX*8 O
      COMPLEX*16,Z,V,X,Y,H,S
      WRITE(*,*)" ENTER CAP VALUES... "
      WRITE(*,*)"      CKO "
      READ(*,*)CKO
      WRITE(*,*)" CK1 AND CK2 "
      READ(*,*)CK1
      WRITE(*,*)"      CK3"
      READ(*,*)CK3
      WRITE(*,*)"      C3 AND C4"
      READ(*,*)C3
      FQ=32000
      T=1/FQ
      CK0=CK0*1E-12
      CK1=CK1*1E-12
      CK2=CK1
      CK3=CK3*1E-12
      C3=C3*1E-12
      C4=C3
      K0=CK0/C3
      K1=CK1/C3
      K3=CK3/C3
      K2=CK2/C4
      OPEN(3,FILE="HOUT.ALL",STATUS="NEW")
C
C     BEGIN LOOP
C
C     RANGE IS 128HZ TO 16.4KHZ
C
C     H(Z)=[CKO/(1+K3)](Z-1)>/[Z**2-Z(2+K3-K1*K2)/
C     (1+K3)+1/(1+K3)]          FIG.5.5-1 (b)     Sanchez
C
```

SUBSTITUTE SHEET

-45-

APPENDIX A   Page 2


```
C
        POWER=6.95
        DO 20 I=1,141
        POWER=POWER+.05
        W=2**POWER
        WT=W*T=6.28
        V=DCMPLX(DBLE(0.),WT)
        Z=CDEXP(V)
C       CALCULATE H(Z)
        X=(Z**2)-Z*(2+K3-K1*K2)/(1+K3)+1/(1+K3)
        Y=KO*(Z-1)/(X*(1+K3))
        O=Y
        IM=AIMAG(O)
        RE=REAL(O)
C
C       CALCULATE GAIN AND PHASE
C
        GAIN=CDABS(Y)
        PHASE=ATAN2(IM,RE)
        DB=ALOG10(GAIN)
        DB=DB*20
C       WRITE(*,*) W,GAIN,PHASE
        WRITE(3,200) INT(W),GAIN
        WRITE(5,200) INT(W),PHASE
        WRITE(4,201) INT(W),GAIN,PHASE
        WRITE(6,200) INT(W),DB
200     FORMAT(1X,I5,10X,F8.3)
201     FORMAT(1X,I5,10X,F8.3,10X,F8.3)
202     FORMAT(1X,F7.2)
20      CONTINUE
        CK0=CKO*1E12
        CK1=CK1*1E12
        CK2=CK2*1E12
        CK3=CK3*1E12
        C3=C3*1E12
        C4=C4*1E12
        WRITE(3,202)CK0
        WRITE(3,202)CK1
        WRITE(3,202)CK2
        WRITE(3,202)CK3
        WRITE(3,202)C3
        WRITE(3,202)C4
        STOP
        END
```

SUBSTITUTE SHEET

APPENDIX B

```
C     THIS PROGRAM SUMS TOGETHER VARIOUS FREQUENCY RESPONSES
C     PREVIOUSLY CALCULATED.  THIS IS TO OBSERVE
C     HOW THE PHASE MAGNITUDE OF VARIOUS PASSBANDS WILL
C     AFFECT EACH OTHER.  ANY COMBINATION OF PREVIOUSLY
C     CALCULATED FREQUENCY RESPONSES CAN BE USED, AS THE
C     PROGRAM PROMPTS THE USER FOR THE NUMBER OF INPUTS
C     AND THE NAME OF EACH.
C
      REAL*4 GAIN(20,141),RE,IM,PHASE(20,141)DPHASE
      REAL*8 WT,W,T,DGAIN,DB
      INTEGER K,L,LQ,FREQ(141)
      COMPLEX*8 O
      COMPLEX*16 Z,V,X,Y,H,S
      CHARACTER*10 IT
      WRITE(*,*)" HOW MANY RESPONSES ARE YOU SUMMING?"
      READ(*,*)K
      OPEN(4,FILE="S.ALL",STATUS="NEW")
      DO 40 I=1,K
      WRITE(*,*)"WHAT INPUT FILE?"
      READ(*,"(A6)")IT
      OPEN(3,FILE=IT,STATUS="OLD")·
      DO 45 L=1,141
      READ(3,*)FREQ(L),GAIN(I,1),PHASE(I,L)
C     WRITE(*,*)FREQ(L),GAIN(I,L),PHASE(I,L)
45    CONTINUE
40    CONTINUE
C     DO THE ADDITION
      DO 60 L=1,141
      RE=0
      IM-0
      DO 55 I=1,K
      RE=RE+GAIN(I,L)*COS(PHASE(I,L))
      IM=IM+GAIN(I,L)*SIN(PHASE(I,L))
55    CONTINUE
      V=DCMPLX(RE,IM)
      DGAIN=COABS(V)
      DPHASE=ATAN2(IM,RE)
      DB=ALOG10(DGAIN)
      DB=DB*20
      WRITE(4,201) FREQ(L),DGAIN,DPHASE
60    CONTINUE
200   FORMAT(1X,I3,10X,F8.3)
201   FORMAT(1X,I5,10X,F.3,10X,F8.3)
      STOP
      END
```

We <u>Claim</u>:

      1.   In a system for fitting a hearing aid
having an adjustable frequency response characteristic
and having a receiver for amplifying sounds in the ear
canal of a client to compensate for a hearing loss, a
method for adjusting the frequency response
characteristic of said hearing aid comprising the
steps of:

      inserting the receiver of said hearing aid
into the ear canal of the client;

      appla signal, having a predetermined
amplitude to an input port of said hearing aid;

      measuring the level of the sound produced in
the ear canal of the client in response to said
signal;

      calculating, from the applied signal and the
measured level of sound produced in the ear canal, a
mechanical frequency response characteristic
representing the interaction of the hearing aid and
the ear canal;

      calculating, from the mechanical frequency
response characteristic and from the hearing loss, a
frequency response characteristic for said hearing aid

- 48 -

to compensate therefor.

          2.   The method set forth in Claim 1 wherein
5   the step of measuring the level of sound produced in
     the ear canal of the client includes the steps of:

          inserting a probe microphone into the ear
     canal of the client; and
10

          calculating the root-mean-square of the
     signal produced by the microphone in response to the
     signal applied to the hearing aid.

15

          3.   In a system for fitting a hearing aid
     having an adjustable amplification factor and having a
     receiver for providing amplified sounds in the ear
     canal of a client to compensate for a known hearing
20   loss, a method for adjusting the amplification factor
     of said hearing aid comprising the steps of:

          a)   inserting the receiver of said hearing
     aid into the ear canal of the client;
25

          b)   adjusting the amplification factor of
     said hearing aid to have a predetermined level

          c)   applying an audio frequency signal,

having a predetermined amplitude, to an input port of
said hearing aid;

       d)   measuring the level of the amplified
5   sound produced in the ear canal of the client in
response to said audio frequency signal;

       e)   calculating, from the applied signal
and the measured level of sound produced in the ear
10   canal, a mechanical frequency response characteristic
representing the interaction of the hearing aid and
the ear canal;

       f)   calculating, from the mechanical
15   frequency response characteristic and from the known
hearing loss, a final amplification factor for said
hearing aid to compensate for said known hearing loss;

       g)   adjusting the amplification factor of
20   said hearing aid to equal said final amplification
factor.

       4.   The method set forth in Claim 3
25   wherein:

       step A further includes the step of
inserting a probe microphone into the ear canal of the
client; and

step D includes the step of generating, from
the signal provided by said probe microphone, an
indication of the sound pressure level produced in the
ear canal of the client in response to said signal.

5

5.   The method set forth in Claim 3 wherein
step F includes a method for determining a maximum
amplification factor which may be used by said hearing
10   aid, said method comprising the steps of:

adjusting the amplification factor of said
hearing aid to a predetermined minimum value;

15   applying a further audio frequency signal to
said hearing aid during a predetermined time interval;

measuring the sound level in the ear canal
of the client after said predetermined time interval
20   and setting an indicator variable in response to said
measured sound level being greater than a threshold
value;

increasing the amplification factor of said
25   hearing aid in response to said indicator variable not
being set;

decreasing the amplification factor of said
hearing aid in response to said indicator variable

_ 51 _

being set; and

     storing said decreased amplification factor
as said maximum amplification factor.

5

     6.   The method set forth in Claim 3 wherein
step F includes the steps of:

10   specifying a target frequency response
characteristic representing a desired level of hearing
for the client fitted with the hearing aid;

     calculating said final amplification factor,
15   such that the combination of said final amplification
factor, said known hearing loss and said mechanical
frequency response characteristic tends to match said
target frequency response characteristic.

20

     7.   The method set forth in Claim 6 wherein
the step of specifying said target frequency response
characteristic includes the steps of:

25   successively positioning a marker on a two-
dimensional graphic display to first, second and
third locations, representing respective first, second
and third points in a graphic representation of said
target frequency response characteristic;

entering coordinates representing said
first, second and third locations in a table;

calculating from the coordinates
5   representing said first, second and third locations, a
portion of said target frequency response
characteristic spanning said first and second points.

10   8.   In a system for fitting a hearing aid
having an adjustable amplification factor having a
receiver for amplifying sounds in the ear canal of a
client to compensate for a hearing loss, a method of
determining a maximum value for said adjustable
15   amplification factor, comprising the steps of:

a)   adjusting the amplification factor of
said hearing aid to a predetermined value;

20   b)   applying an audio frequency signal to
said hearing aid during a predetermined time interval;

c)   measuring the sound level in the ear
25   canal of the client after said predetermined time
interval and setting an indicator variable in response
to said measured sound level being greater than a
threshold value;

30   d)   increasing the amplification factor of

said hearing aid in response to said indicator
variable not being set;

   e)   decreasing the amplification factor of
said hearing aid in response to said indicator
variable being set;

   f)   storing said resultant amplification
factor as said maximum value.

   9.   The method set forth in Claim 8
wherein:

   step D further includes the step of
repeating steps B through D while said amplification
factor is less than a predetermined maximum possible
value and until said indicator variable is set after
step C; and

   step E further includes the steps of
resetting said indicator variable and repeating steps
B, through E until said indicator variable is not set
after step C.

FIG. I

FIG. 2B

FIG. 2A

FIG. 3

TO/FROM
CHEST PACK

322   324   323

ADC   RMS
       CONV   MUX   MIC_L
                       MIC_R

320   321

SIGNAL
GENERATOR   DE-
             MUX   SIG_L

                    SIG_R

TO/FROM
COMPUTER   310   312   316   318   CK_PA

RECEIVER
XMIT   μP   S/R   SYNC.
               I/O   PA

                      L/R

                      MUTE

INTERFACE   UNIT   18   BURN

                                        424

MIC_L   AMP

MIC_R   422   AMP

SIG_L

SIG_R

MUTE   LEVEL   420
                           SHIFT

L/R   412   416

                                  414   DE-   L
                           LEVEL   MUX
PA   CK_PA   S/R   SHIFT

                                         R

                           418
                           LEVEL
                           SHIFT

BURN   LEVEL   BURN
                                  SHIFT
CHEST   PACK   20   419

TO/FROM
INTERFACE   FIG. 4   TO/FROM
UNIT   HEARING AIDS

FIG. 3A

FIG. 3A

350 — WAIT FOR NEXT COMMAND

380 — RECEIVE SIGNAL FROM PROBE MICROPHONE REQUEST PARAMETERS FROM COMPUTER

352 — COMMAND TYPE

354 — CHANGE AMPLIFICATION FACTOR REQUEST PARAMETERS FROM COMPUTER

382 — SELECT HEARING AID

370 — SEND SIGNAL TO HEARING AID REQUEST PARAMETERS FROM COMPUTER

355 — MUTE SELECTED HEARING AID

384 — DIGITIZE SIGNAL FROM SELECTED HEARING AID

372 — CONDITION SIGNAL GENERATOR TO PROVIDE REQUESTED SIGNAL

356 — LOAD SHIFT REGISTER WITH NEW AMPLIFICATION FACTORS

386 — FORMAT DATA

374 — SEND TO SELECTED HEARING AID

358 — SEND TO SELECTED HEARING AID

388 — SEND TO COMPUTER

360 — RECEIVE CHECK DATA

362 — ERROR

364 — REVERSE MUTE

FIG. 5-1

FIG. 5-2

FIG. 6

```
          ┌──────────────────┐
          │    CALIBRATE     │
          │   HEARING AID    │ ⟋ 710
          └──────────────────┘
                    │
                    ▼
          ┌──────────────────┐
          │    DETERMINE     │
          │     FEEDBACK     │
          │     CEILING      │⟋712
          └──────────────────┘
                    │
                    ▼
          ┌──────────────────┐
          │   SPECIFY LOSS   │⟋714
          │      CURVE       │
          └──────────────────┘
                    │
                    ▼
          ┌──────────────────┐
          │  SPECIFY TARGET  │⟋716
          │      CURVE       │
          └──────────────────┘
                    │
                    ▼
          ┌──────────────────┐
          │ ADJUST CHANNEL   │
          │  AMPLIFICATION   │
          │  FACTOR TO FIT   │⟋718
          │  TARGET CURVE    │
          └──────────────────┘
                    │
                    ▼          ⟋720            ⟋724
                ◇─────────◇          ┌──────────────┐
               ╱  FURTHER  ╲────────▶│ SAVE CLIENT  │
               ╲ ADJUSTMENT ╱        │    DATA      │
                ◇─────────◇          └──────────────┘
                    │
                    ▼          ⟋722
          ┌──────────────────┐
          │  ADJUST  TARGET  │
          │      CURVE       │
          └──────────────────┘
```
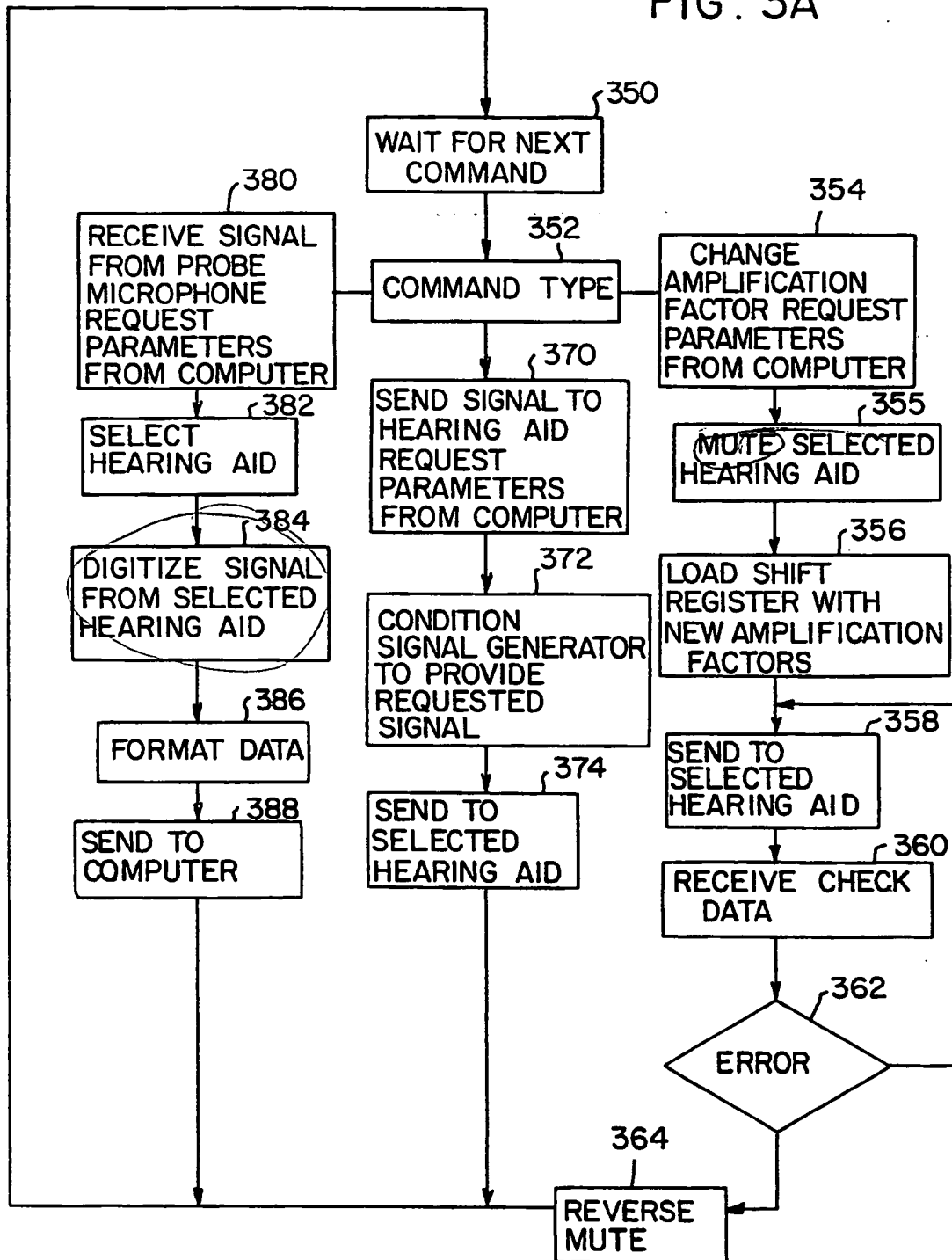
FIG. 7

FIG. 8

SET ALL ATTENUATORS
TO A KNOWN LEVEL   — 810

CHANNEL = 0   — 812

CHANNEL = CHANNEL + 1
SELECT CHANNEL   — 814

APPLY SOUND SIGNAL
TO SELECTED CHANNEL   — 816

GET SOUND PRESSURE
IN EAR CANAL   — 818

CALCULATE
AMPLIFICATION FACTOR
FOR CHANNEL   — 820

STORE AMPLIFICATION
FACTOR   — 822

826

END   CHANNEL = 13   — 824

SET ALL ATTENUATORS
TO MAXIMUM
ATTENUATION — 910

↓

CHANNEL = 0 — 912

↓

CHANNEL = CHANNEL + I — 914

↓

DECREMENT
ATTENUATION — 916
LEVEL

↓

EXCITE SYSTEM
WITH NOISE IN
FREQUENCY BAND — 918
OF SELECTED CHANNEL

↓

MEASURE RESPONSE — 920
AFTER EXCITATION

↓

SIGNAL
PRESENT ? — 922

924
INCREMENT
ATTENUATION
LEVEL

ATTENUATION
AT LOWEST
LEVEL ?
926

928
MARK GAIN AS
MAXIMUM FOR
CHANNEL ;

SET ATTENUATION
LEVEL TO MAX.

FIG. 9

↓ 930

CHANNEL = 13

↓ 932

END

GENERATE TABLE OF
NOMINAL AMPLIFICATION
FACTORS FOR EACH
CHANNEL ⌐1010

ADD CALIBRATION
DATA TO TABLE ⌐1012

LIMIT TABLE ENTRIES
TO MAXIMUM VALUES
DETERMINED BY
FEEDBACK ANALYSIS ⌐1014

FOR EACH CENTER
FREQUENCY CALCULATE
THE DIFFERENCE
BETWEEN LOSS CURVE
AND TARGET CURVE ⌐1016

FOR EACH CENTER
FREQUENCY IF
DIFFERENCE > MAX AMPL.
FACT. SUBSTITUTE MAX
VALVE IF DIFFERENCE <
MIN AMPL. FACT.
SUBSTITUTE MIN. VALUE ⌐1018

SEARCH EACH COLUMN
OF TABLE FOR AMPL.
FACT. THAT MATCHES
THE DIFFERENCE
VALUE FOR THE
CORRESPONDING
CHANNEL ⌐1020

CALCULATE ACTUAL
FREQUENCY RESPONSE
CHARACTERISTIC OF
TOTAL SYSTEM ⌐1022

ADD CALCULATED
FREQUENCY RESPONSE
CHARACTERISTIC TO
LOSS CURVE ⌐1023

SUBTRACT ACTUAL
AMPLIFICATION FACTORS
FOR 13 CHANNELS FROM
TARGET CURVE ⌐1024

SUBTRACT CURRENT
ERROR VALUES FROM
PRIOR ERROR VALUES
TO GENERATE ERROR
DIFFERENCE VALUES ⌐1026

ERROR
DIFFERENCE LESS
THAN THRES-
HOLD ⌐1027

Y → DONE ⌐10

N

SUBTRACT CURRENT ERROR
VALUES FROM
AMPLIFICATION FACTOR
DIFFERENCE VALUES ⌐1028

FIG. 10

FIG. 11

INTERNATIONAL SEARCH REPORT

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all)[6]

According to International Patent Classification (IPC) or to both National Classification and IPC

IPC5: A 61 B 5/12, H 04 R 25/00

II. FIELDS SEARCHED

Minimum Documentation Searched[7]	
Classification System	Classification Symbols
IPC5	H 04 R, A 61 B

Documentation Searched other than Minimum Documentation to the Extent that such Documents are Included in Fields Searched[8]

III. DOCUMENTS CONSIDERED TO BE RELEVANT[9]

Category[*]	Citation of Document,[11] with indication, where appropriate, of the relevant passages[12]	Relevant to Claim No.[13]
A	US, A, 4596902 (S. GILMAN) 24 June 1986, see column 2, line 29 - line 68; abstract --	1-9
A	US, A, 4289143 (F. CANAVESIO ET AL) 15 September 1981, see the whole document --	1-9
A	DE, C2, 3113849 (SIEMENS AG) 11 September 1986, see the whole document --	1-9

* Special categories of cited documents: [10]

"A" document defining the general state of the art which is not considered to be of particular relevance

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"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

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"X" document of particular relevance, the claimed invention cannot be considered novel or cannot be considered to involve an inventive step

"Y" document of particular relevance, the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

IV. CERTIFICATION

Date of the Actual Completion of the International Search	Date of Mailing of this International Search Report
28th May 1990	21. 06. 90

International Searching Authority	Signature of Authorized Officer
EUROPEAN PATENT OFFICE	M. Peis [M. PEIS]

This annex lists the patent family members relating to the patent documents cited in the above-mentioned international search report.
The members are as contained in the European Patent Office EDP file on   07/05/90
The European Patent office is in no way liable for theseparticulars which are merely given for the purpose of information.

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US-A- 4596902	24/06/86	NONE	
US-A- 4289143	15/09/81	EP-A-B- 0014324	20/08/80
DE-C2- 3113849	11/09/86	NONE	